THERMAL NOISE MODEL OF CAPACITIVE ACTIVE ELECTRODE FOR INDIRECT-CONTACT ECG

Yong Gyu Lim
Department of Oriental Biomedical Engineering, Sangji University, Wonju, Republic of Korea

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Abstract: The indirect-contact ECG (IDC-ECG) shows large background noise in comparison with conventional ECG measurement. To improve the signal quality, close study of the background noise is necessary. This study was carried out to investigate how much the thermal noise influences the background noise in IDC-ECG. To do so, the thermal noise model was built for the active electrode. And then, the parameters which determine the thermal noise were estimated by measuring the gain of the active electrode. Finally, the level of thermal noise was estimated and compared with actual background noise. The results show that the thermal noise is the dominant component of background noise and the intrinsic noise of the preamp's active devices is negligible.

1 INTRODUCTION

The need for physiological signal measurement at home is increasing due to the interest in the early disease detection and the continuous or frequent prognosis monitoring. In addition, the need for health monitoring in daily life is increasing for the sake of improving life quality. For the above stated uses, we introduced an ECG measurement method adequate for daily long-term measurement (Lim, 2006). Using the introduced method, ECG was measured through usual clothes without direct contact between the skin and instrument.

In comparison with conventional ECG, the background noise of IDC-ECG is large, and its characteristics vary widely according to the measurement conditions. To improve the signal quality, an analysis of the background noise is necessary.

In this study, we carried out an experiment to determine the gain of IDC-ECG measurement for some sample cloth, cotton. And based on the measured gain, the thermal noise level was estimated. By the comparison between the estimated thermal noise and the actual background noise of IDC-ECG, we investigated how much the thermal noise influences the background noise.

2 METHODOLOGY

2.1 Indirect-contact ECG Measurement

Indirect-contact ECG (IDC-ECG) is a method that enables ECG measures without direct contact between the electrodes and bare skin. The method is composed of two components. One is a high-input-impedance capacitive and active electrode that enables ECG measurement through the high impedance of the clothes. The other component is the indirect-contact grounding through the clothes.

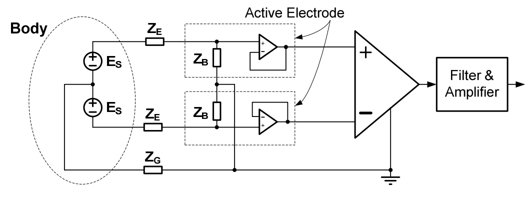

Figure 1: Diagram of the IDC-ECG measurement system showing the electrode impedance (Z_{CLTH}), input impedance of the preamp (Z_B), and impedance between the body and ground (Z_G).

Figure 1 shows the diagram of the whole measurement system. The ECG generated in the subject's body is sensed by the two active electrodes through cloth impedance Z_E. The difference between the outputs of the two electrodes is acquired by an

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instrumentation amplifier, and it is filtered and amplified by circuitry behind the instrumentation amplifier. The final bandwidth is 0.5-35 Hz and the total gain is 5000.

2.1.1 Frequency Response of the Active Electrode

Figure 2 shows a model of the presented active electrode. The preamp in the active electrode is simply composed of one op-amp and one discrete resistor (R_B), which is a path for the bias current of the op-amp's input terminal. C_B is the parasitic capacitance between the input and the ground. The parallel connection of R_A and C_A represents the input impedance of the op-amp. Z_E is the electrode impedance, which is present between the electrode face and the skin. The electrode impedance is represented as a parallel connection of R_{CLTH} and C_{CLTH}.

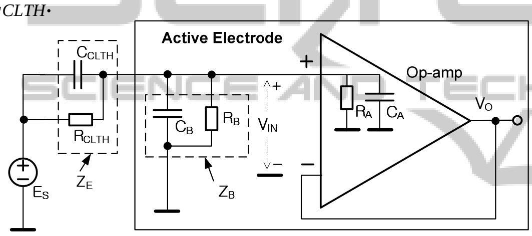

Figure 2: Model of active electrode with signal source.

The ECG source signal (the potential variation at skin surface) is represented as a voltage source E_S. Therefore, the gain for the input E_S is

$$G_S(s) = \frac{Z_B // Z_A}{Z_E + Z_B // Z_A} \tag{1}$$

In the electrode designed in this study, the input impedance Z_A of the op-amp is so large in comparison with Z_B that it can be disregarded. Therefore, the gain leads to

$$G_S(s) = \frac{R_B + C_{CLTH}\, R_B\, R_{CLTH}\, s}{(R_B + R_{CLTH}) + (C_B + C_{CLTH}) R_B\, R_{CLTH}\, s} \tag{2}$$

2.1.2 Model of the Thermal Noise in the Electrode

Figure 3 shows various noise sources to be considered for the evaluation of the electrode output noise. E_V and I_A denote the voltage noise source and the current noise source, respectively, of the intrinsic op-amp noise (Ott, 1988). We can obtain detailed information about the op-amp noise from the data sheet provided by the manufacture (TI, 1998). E_{CLTH} and E_B are thermal noises (also called as Johnson noises) produced at the resistances R_{CLTH} and R_B, respectively. The thermal noise amplitude is described by the root-mean-square (rms) voltage as follows:

$$V_{rms} = \sqrt{4kTBR} \tag{3}$$

where k: Boltzmann's constant (1.38 x 10^{-23} joules/ $^\circ$ K)
T: absolute temperature ($^\circ$ K)
B: bandwidth of noise (Hz)
R: resistance (Ω).

The thermal and op-amp noises are generated at the active electrode components, and can be called internal noises

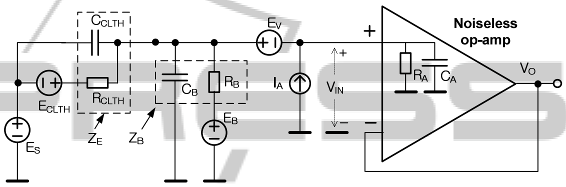

Figure 3: Model of active electrode including noises.

The effects of noises on the op-amp output can be compared with each other by conversion to the voltages observed at the op-amp input. The total voltage V_{IN} at the op-amp input is expressed as

$$V_{IN}(f) = G_S(f) E_S(f) + V_N(f) \tag{4}$$

where G_S is the ECG signal gain as defined in eqs. (1) and (2), and V_N is the total noise shown at the op-amp input.

It seems reasonable to suppose that the noises are random and uncorrelated with each other (Ott, 1988). On this assumption, the total noise voltage can be decomposed as

$$V_N(f) = \left(V_V(f)^2 + V_A(f)^2 + V_B(f)^2 + V_E(f)^2 + V_{EXT}(f)^2\right)^{1/2} \tag{5}$$

where the total noise $V_N(f)$ and all of its components, such as $V_B(f)$, are RMS voltages per square root of bandwidth and their unit is $\mathrm{V}/\sqrt{\mathrm{Hz}}$.

The voltage components at the op-amp input are acquired as shown below.

$$V_V(f) = E_A(f) \tag{6}$$

$$V_A(f) = \frac{R_{CLTH} R_B}{D} I_A(f) \tag{7}$$

$$V_B(f) = \frac{R_{CLTH}}{D} E_B(f) \tag{8}$$

$$V_E(f) = \frac{R_B}{D} E_{CLTH}(f) \tag{9}$$

$$V_{EXT}(f) = \frac{2\pi f C_{EXT} R_{CLTH} R_B}{D} E_{EXT}(f) \qquad (10)$$

where

$$D = \left| R_{CLTH} + R_B + j\,2\pi f\, R_{CLTH} R_B (C_{CLTH} + C_B + C_{EXT}) \right|$$

Since the input impedance of the op-amp was much greater than Z_B in the case of our electrode design, we disregard it for the induction of equations (6) to (10).

3 RESULTS

Equation (2) shows that the cloth impedance (Z_E) can be estimated if we know the gain and input impedance of the preamp (Z_B). The Z_B is already known because it is determined by the design (R_B = 3 GΩ, C_B = 17 pF). The gain was measured by the experimental setup shown in fig. 4. The electrode was laid on a sample cotton cloth and sinusoidal signal was provided by a copper plate put under the cloth.

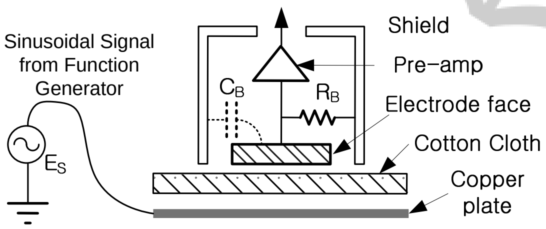

Figure 4: Gain measurement setup.

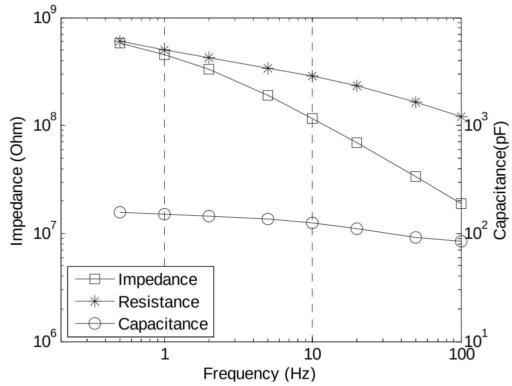

Figure 5: Estimated electrode impedance (Z_E), and its parallel component R_{CLTH} and C_{CLTH}.

Figure 5 shows the estimated electrode impedance which was obtained by applying (2) to the gain measurement.

We can calculate the internal noise components by applying eqs. (6) - (10) to the estimated Z_E shown in Fig. 5. Figure 6 shows the estimated internal noise components for the sample cloth. The figure shows that the noise spectral density decreases at high frequency range that is expected by (6)-(10). In the graph, we can also see that the noise components (V_V and V_A), originating from the intrinsic op-amp noise, are negligible in comparison with the other two thermal noise components (V_B and V_E).

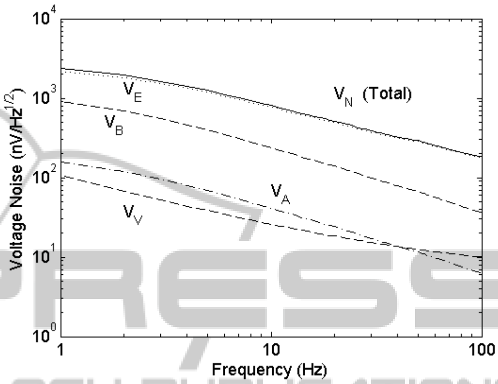

Figure 6: Noise spectral densities were estimated theoretically for the cotton sample cloth. Each noise component is defined in (6) – (10).

It is desirable to compare the spectral density of background noise in actual ECG with that of theoretical thermal noise. However, we cannot get the spectral density of background noise alone because the signal ECG cannot be removed. Instead of the quantitative comparison, a comparison between waveforms by eye was performed. For that comparison, pseudo-noise waveform was generated. Its spectral density was $\sqrt{2}$ times the spectral density shown in Fig. 6, because the ECG was measured through two electrodes.

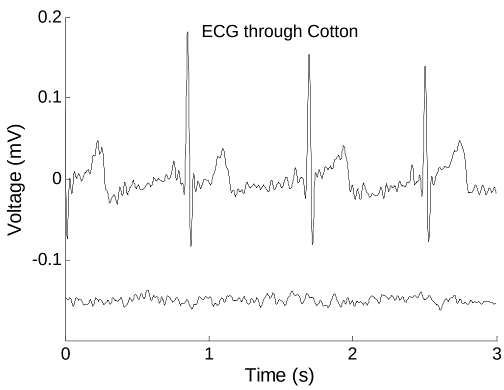

Figure 7: ECG waveforms acquired by IDC-ECG with Ag-AgCl grounding (upper trace) and pseudo-noises synthesized artificially for cotton (lower trace). The frequency band was from 1 to 35 Hz.

Figure 7 shows the ECG and its corresponding synthesized pseudo-noise for the sample cloth. The figure shows that the background noise of actual ECG and pseudo-noise look similar to each other in amplitude and morphology. This result shows that the thermal noise is the dominant component of IDC-ECG.

4 CONCLUSIONS

A thermal noise model for the active electrode was built, regarding the impedance between the electrode and body through clothes as parallel connection of a resistance and a capacitance. The results show that the thermal noise generated in the resistances of the clothes and the electrode is the dominant component of the background noise in the IDC-ECG. And furthermore, the intrinsic noise of the preamp's active devices is negligible in comparison with the thermal noise generated by the passive components. This study explains why the IDC-ECG makes so large background noise in comparison with conventional ECG measurements and shows what to do in order to reduce the background level.

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